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To Predict the Body's Strength

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Although the active and passive strength of the human body has been of interest for many medical and ergonomic problems, it was the emergence of aerospace medicine which required the fundamental and practical studies that provide today's body of data in this field. After the early pioneering years of the 1930's, 1940's and 1950's, the last 25 years brought a maturing of the field of biodynamics, its methods, tools, and theoretical foundations. This overview discusses some of these advances, their contributions to aerospace safety, as well as their applications to the broader areas of traffic safety, orthopedic biodynamics, medicine and ergonomics. To meet future aviation as well as societal challenges, steady efforts by a few centers of excellence are required to integrate operational, experimental, and theoretical advances into sophisticated prediction capabilities.

AS PART OF THE Aerospace Medical Division's 25th Anniversary Celebration on 6-7 October 1986, there was a "Symposium of Firsts" commemorating great milestones of aerospace medical history. The "Firsts" would not be there without the many small steps which preceded the first giant step. They would be meaningless and forgotten without the subsequent steps of advancing, broadening, and applying the results in the daily, nitty-gritty work of many people. The early, sustained acceleration tolerance work during and after World War II and the pioneering, acceleration/deceleration experiments in the 1950's and 1960's have been recently reviewed and are well documented (23,26). Is there much to add about the body's resistance to mechanical forces? Almost 20 years ago at the Sixth Series of the USAF School of Aerospace Medicine's Lectures in Aerospace Medicine, I reviewed the field and concluded that it had reached a certain maturity and cohesiveness (25). I will review here some of the recent advances and their usefulness to applications.

Biodynamics was hardly a term, let alone a discipline, before aviation medicine needed it. The body's strength and resistance to forces was of primary interest with respect to the internal muscle forces, their application to the skeleton, and the net forces available for work and locomotion. Artists and anatomists provided primarily the few data available until the beginning of World War II (17) (Fig. 1).

A systematic measurement program of the muscle

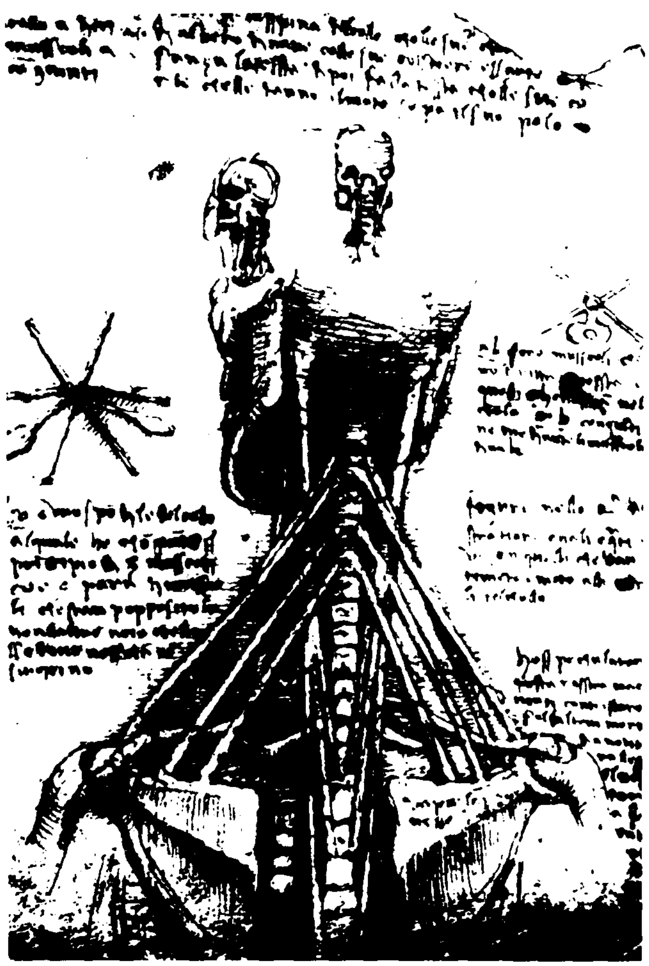

Fig. 1. The anatomy of the human neck. Drawing by Leonardo Da Vinci (17).

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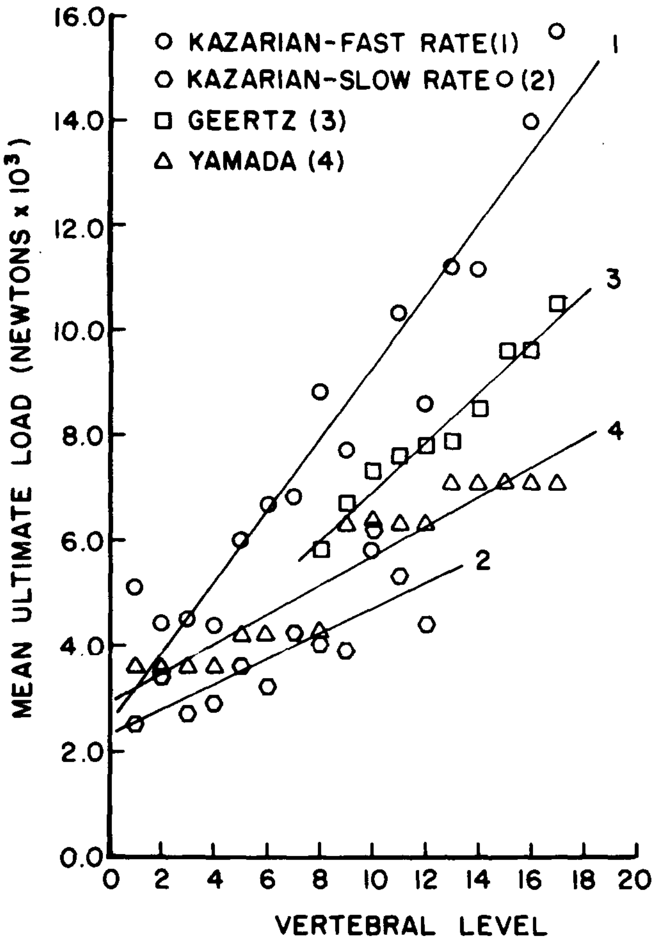


Fig. 2. Examples of the dependence of ultimate load strength of human vertebral bodies on loading rate and the vertebral level. The data show the increase in vertebral body strength between T_1 (level 1) and L_5 (level 17) (12).

strength of male and female subjects of different body sizes and weights and for various working positions was recently conducted at our laboratory because of the increasing necessity of these measurements for efficient ergonomic design and crew selection (15,16). The real impetus to the field came when the forces that potentially could be applied to the body increased manyfold over the forces involved in daily, human activity, for which the body had evolved.

Parachute opening shock, and crash protection raised the question of the body's ultimate strength limits. While the first experiments to probe human tolerance to impact forces had just been started at novel test facilities of the late 1930's, the idea of emergency escape from disabled aircraft emerged. This idea generated the question, "How much acceleration in the buttocks-to-head direction is tolerable in ejecting a crewmember from the aircraft?" At that time (1938), it took just 24 hours of experimentation at the German deceleration facility to determine that the required 10-12 G headwards acceleration would be tolerable for 1 second (21). And that was the criteria for the first ejection seat ever built, which led to the production of about 1,000 seats by the end of the war and approximately 60 successful ejections; that was almost 50 years ago.

What have we learned since, and why is further research in biodynamics indicated? We had automobiles 50 years ago, and protecting their occupants was not a problem. But even today, with all our knowledge and advanced technology, mechanical injury remains one of our main diseases, according to the Committee on Trauma Research of the National Research Council (NRC) and the Institute of Medicine in their report, "Injury in America." Congress recently directed the Centers for Disease Control to start a large new program on trauma research and injury prevention (4). This major new program—to be undertaken without participation of the DOD, which up to now did most of the

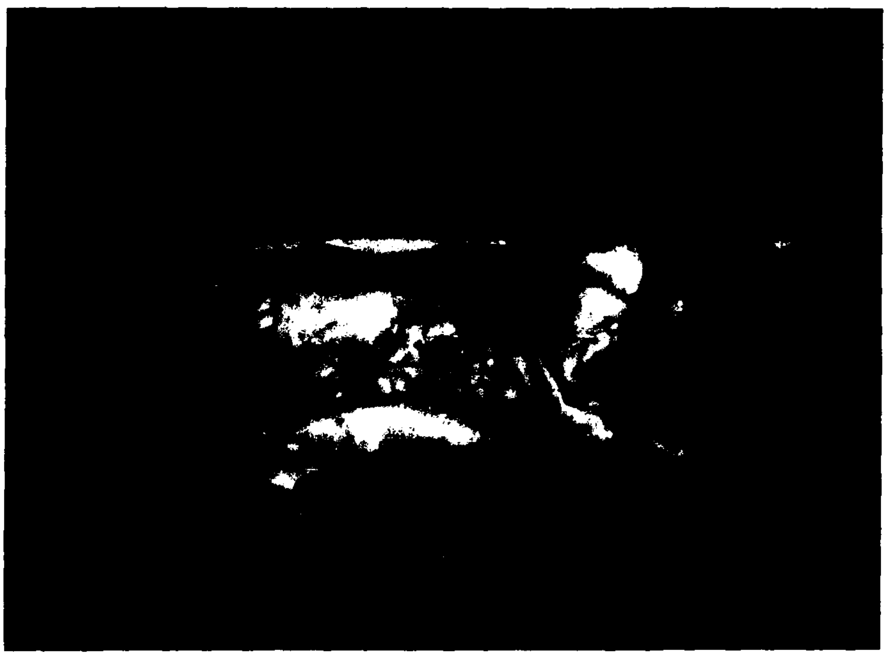


Fig. 3. Compression response of a vertebral body (strain rate 2,100 in·min⁻¹). The bone marrow in the vertebral centrum is ejected by the hydraulic pressure through orifices in the centrum. The bone recovers remarkably from the compressive deformation. At still higher strain rates the fluid cannot escape and the bone splinters under the high pressure (11).

human impact research for the military and civilian sector—highlights the continuing need for biodynamic research.

Asking, "Why more research, or have not most problems been solved?" is like asking, "Why build new aircraft or a super cockpit?" since both the airplane as well as a cockpit are available today. Taking the first steps into virgin territory poses enormous problems, but the findings can represent giant steps, and the relative payoff is high. However, the real gain might only be realized after many more steps, successes, and failures that represent day after day and year after year, consequential work which gives us the capabilities that the first step allowed us to envision. In biodynamics, this goal is to physically and mathematically describe the human body, so we can predict its behavior under internal or external mechanical forces. Using description to predict behavior is the way we learned to calculate the response of the physical world.

Biodynamics has expanded into an interdisciplinary field during the last 50 years, feeding into and supporting many areas of physiology, medicine, human factors engineering, and ergonomics. Biodynamics' major boost to this growth came from beginnings in aerospace medicine which are hardly ever traced to today's mature impressive results. The first textbooks in this field are just now being written (7), and I could do no justice to the technical progress in a short review. Let me instead illustrate with a few examples and summaries what our laboratory has accomplished, where we stand with respect to basic knowledge, and where and how this knowledge is being applied.

To understand the reason for the body's complex response to mechanical forces and to explain the observed trauma, it is necessary to characterize the material properties of the various components and substructures. This characterization by itself developed into a specialized field, which borrowed and developed its own tools and methods (5,27). We are still far from being able to tabulate the basic physical and strength properties of all body tissues, not to mention the added difficulty that these properties change markedly when

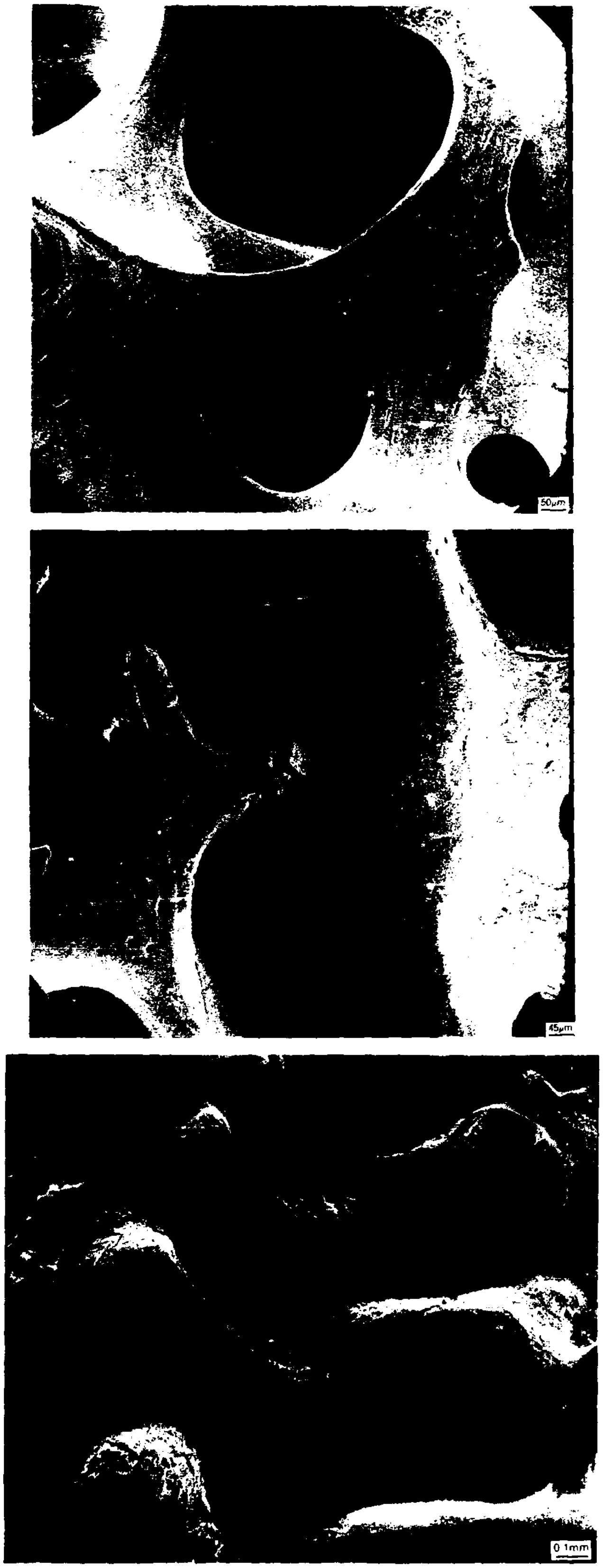


Fig. 4. Scanning electron microscopic (SEM) views of trabeculae from vertebral bodies compressed at 8.89×10^{-5} m·s^{-1}: (upper) normal uncompressed trabeculae in the end zone; (center) a fractured horizontal trabeculae; (lower) S-shaped bending of long trabeculae of the central zone (6).

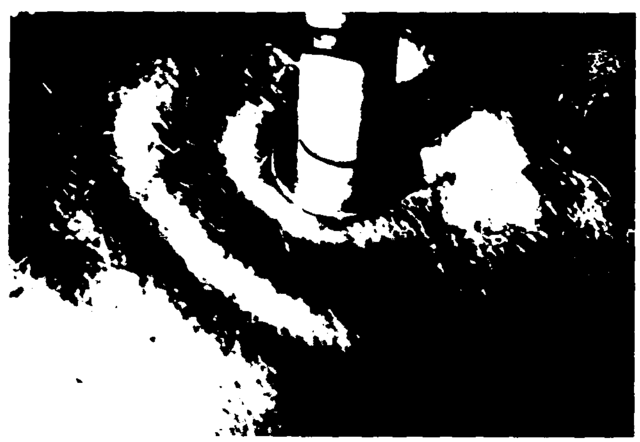


Fig. 5. Surface waves on the human thigh in stroboscopic illumination (excitation frequency 68 Hz; diameter of vibrating piston 2.6 cm; wave velocity 160 cm · s^{-1}). This observation allows estimation of the shear elasticity of soft tissue *in vivo* (24).

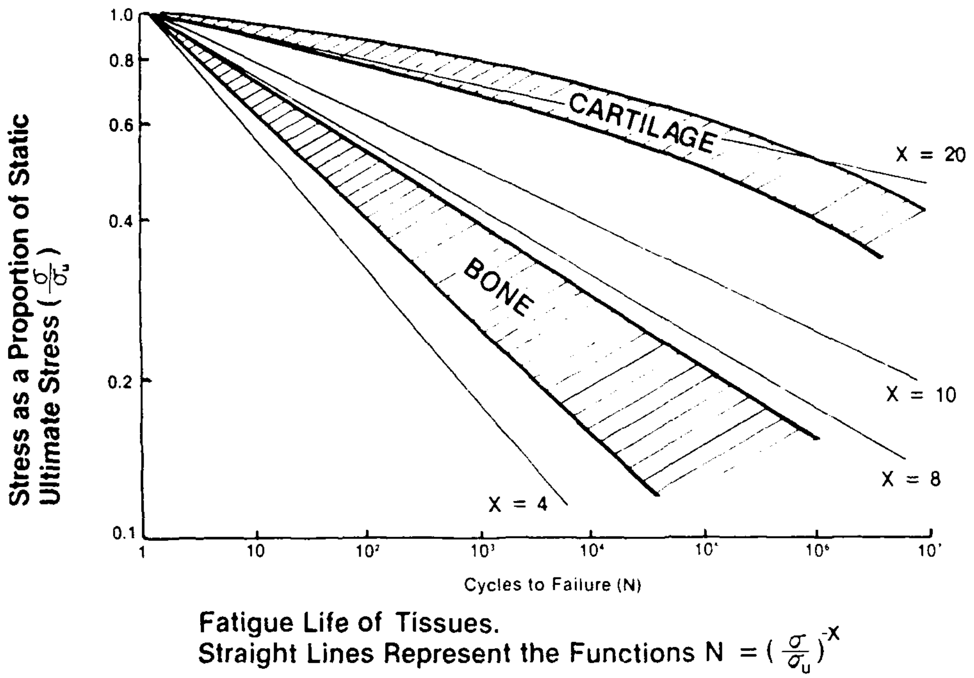


Fig. 6. Fatigue life of hard and soft tissues. The cycles to failure N are shown as a function of the ratio of the applied dynamic stress to the static ultimate stress (22).

not measured *in vivo* and that they can change with stature, age, and physical condition. Researchers interested in aircraft ejection problems concentrated investigative efforts on the spinal column, an unbelievably complicated, composite materials structure (11,13). Fig. 2 illustrates the dependence of ultimate strength on vertebral level and strain rate. Despite the work of many laboratories, we still have not completely collected the material characteristics of all bones of the human skeleton (12). The nonlinear strength characteristics of the vertebral bodies are of particular interest in connection with ejection-seat injuries. These injuries are caused by the complicated hydrodynamic phenomena occurring during severe compression and by the vertebral bodies' complete change when going from slow to rapid compression (Fig. 3). In spite of the compression observed (50% of original height), the boney vertebral structure expands again following release to almost its original height (75%) and even microscopic examination might indicate that no serious injury (i.e., fracture) occurred. However, electron-microscopic (EM) analysis reveals the microtrauma in the trabecular structure and the inherently changed properties of the vertebra (Fig. 4).

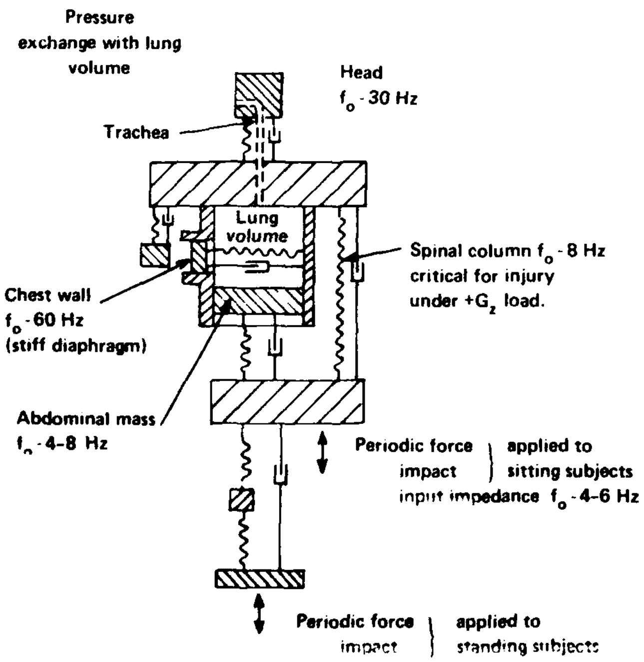


Fig. 7. Lumped parameter model for calculating body deformation and some physiological and subjective responses of the human body exposed to longitudinal G_z vibration. The approximate resonance frequencies of the various subsystems are designated by f_o (25).

Similar complexity confronts us when we try to measure the strength of soft tissues and tendons. Forty years ago we thought it would be enough to determine the soft tissue's properties from its response to vibratory excitation (24). For example, the wave pattern on the human skin (Fig. 5) allowed us to calculate, among other quantities, the tissue's shear elasticity and viscosity. These data were useful to explain hemodynamic and psychophysical phenomena. They do not help us understand the changes of the parameters with muscle tone or the tearing of soft tissue under stress. We had to use microscopic cytodynamic investigations. We would expect the repeated exposure to vibratory loads or shock loads to lead to a dependence of soft tissue failure load on the repetition cycle as illustrated in Fig. 6 (22). Unfortunately, the ranges shown have only been determined approximately, although they have clear implications for the biological, mechanical, and fatigue life of a human under stresses as they occur in industrial and military situations.

Knowing tissue properties, body shape, and composition should allow us to calculate the body deformation under external forces, determine stress concentrations, and determine the loads under which various components fail (i.e., exhibit injury). Unfortunately, the system is not so simple. It took us many years to recognize

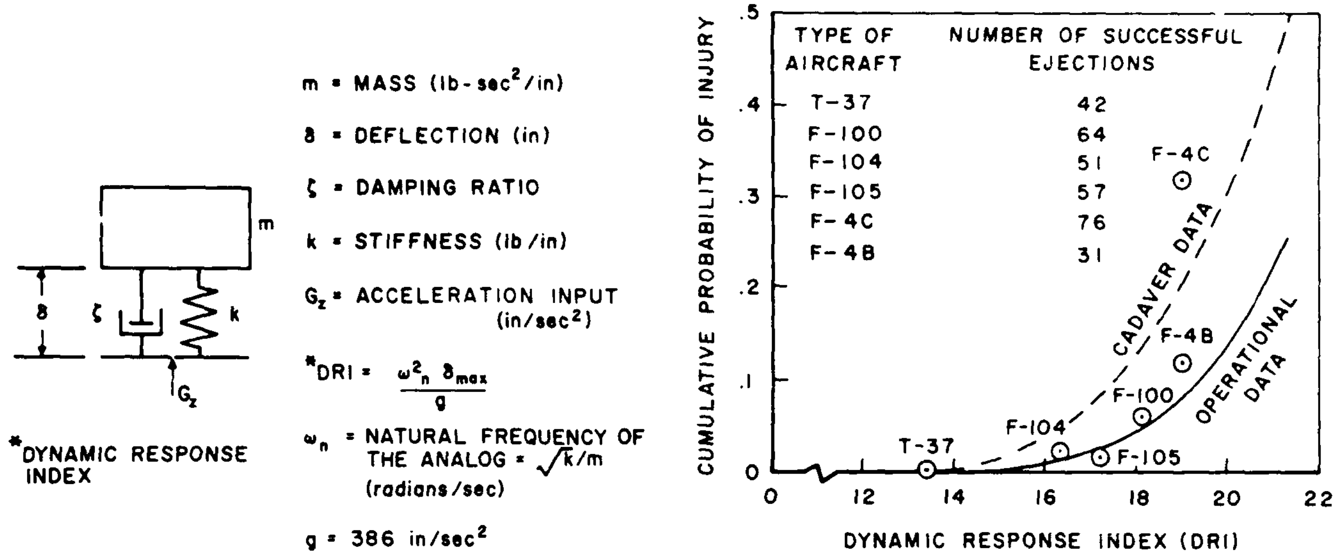


Fig. 8. Application of the DRI to predict probability of spinal injury from seat ejection. (Left) The DRI lumped parameter model. The mass, m, represents the upper torso, the stiffness, K, the spinal spring. Injury probability is expressed in terms of maximum spinal compression δ under the impact acceleration Gz. (Right) Operational ejection injury data compared to DRI prediction (2,18).

and describe the mobility of the various subsystems of the body such as the head, the upper torso, and the abdominal viscera (25). These studies, conducted primarily on human volunteers on vibration tables, were originated to study human sensitivity and tolerance to vibration stresses.

The manyfold symptoms observed began to make sense and to fit into a describable pattern only when we recognized their correlation with maximum tissue stresses caused by relative displacement of body segments or subsystems under the vibratory load. For example, the abdominal viscera vibrated with respect to the body's skeletal structure and had a resonance with maximum tissue stress at a specific frequency, the same frequency at which subjective symptoms (and injury in animal experiments) were most pronounced. To explain

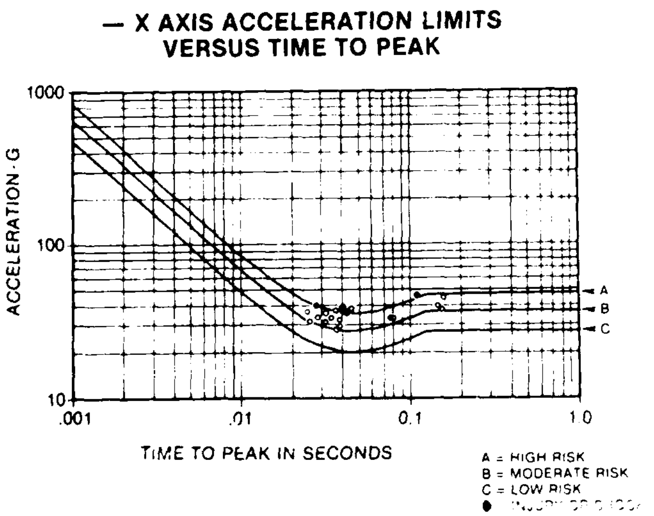


Fig. 9. Example of 6 degree-of-freedom impact acceleration limits used for advanced escape systems: Injury risk levels for −Gx half-sine acceleration pulses. The data points represent human exposures. The general shape of the curves are supported by additional subjective and physical response tests with volunteer subjects at lower exposure levels (3).

and describe these findings, the first lumped-parameter mathematical models were developed. They simulated the human body by masses and springs and provided a gross picture of its deformation under external forces (25) (Fig. 7). Based on the work by Latham on body ballistics (14) and the work in our laboratory and others, these models were reduced to a simple injury model to predict spinal injury probability as a function of the ejection seat acceleration profile.

Although the influence of the dynamics of the spine on the tolerability of various ejection profiles had already been recognized and addressed in Ruff's early work, it was not until the "Dynamic Response Index" (DRI) was formalized, standardized, and used to compare operational injury data with predicted severity that it was possible to understand the differences between different seat designs (Fig. 8) (2,18). The severity of spinal injury corresponds in the DRI to the degree of longitudinal (Z-axis) compression of the single spring representing the spine. The differences in location and type of injury along the spinal column cannot be predicted by the DRI (Fig. 8). Although many more sophisticated, lumped-parameter spinal models and even nonlinear spinal models were proposed in the meantime (18), the simplicity and reasonable accuracy achieved with the DRI concept were the factors which kept it our standardized design tool for escape-system development for more than 15 years.

Only recently has the DRI concept been reanalyzed and expanded. When refined, not one but six degrees-of-freedom (DOF) acceleration exposure limits were needed for the advanced crew escape system technology (CREST) program (8). The control system steering the ejection seat and the life-threat assessment system, on which the escape sequence and trajectory are based, need continuous injury risk assessment for the potential accelerations applied to the seat. Brinkley (3) developed a DRI model for each orthogonal axis and defined for each axis the injury risk curves corresponding to the 50, 5, and 0.5% spinal injury probability for the Z-axis. Al-

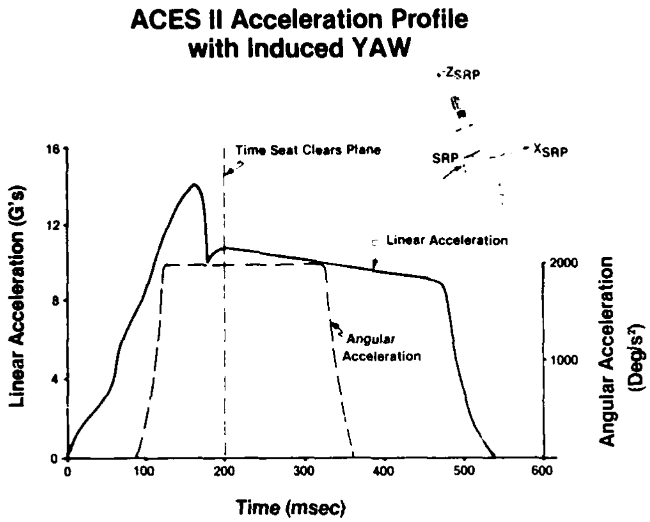
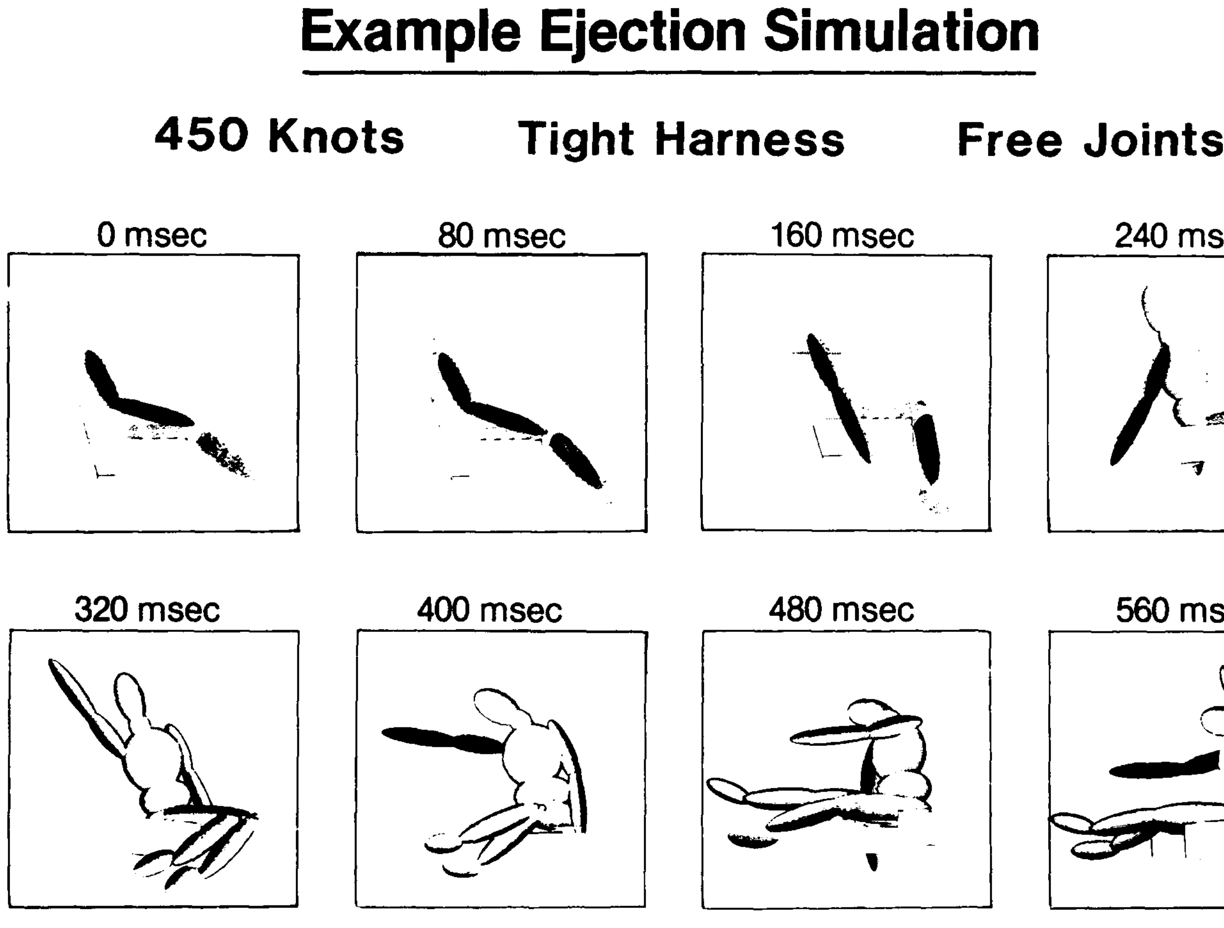


Fig. 10. Ejection simulation with the ATBM (9): (Left) the linear and angular accelerations [coordinate system originates at seat reference point (SRP)] imparted on ACES II ejection seat. (Below) body motions as a result of ejection into a 450-kn windstream with the acceleration presented in upper figure.



though documented injury data for the ±X and ±Y axes are sparse, and the moderate and low-risk levels had to be based on several assumptions and interpolations, the uniform, six degree-of-freedom approach is a major step forward. It provides the necessary guidance for the escape system development by defining curves of equal tolerability for all degrees of freedom. An example for the X-axis is presented in Fig. 9. The angular acceleration exposure limits are based with reasonable justification on the effects of their translational acceleration components. The theoretical shape of the curves, the natural frequency of the DRI model (which is different for each axis), has been confirmed by recent volunteer tests. No United States test facilities exist to verify the limits for rotational acceleration, a shortcoming which might have to be considered in future facility plans.

Comparing these six DOF exposure limits with the tolerance curves available for designers in the 1950's and 1960's, our practical progress is evident. However, in research our progress has gone much farther. The two more detailed, advanced models developed allow us to calculate, at least in the laboratory though not yet

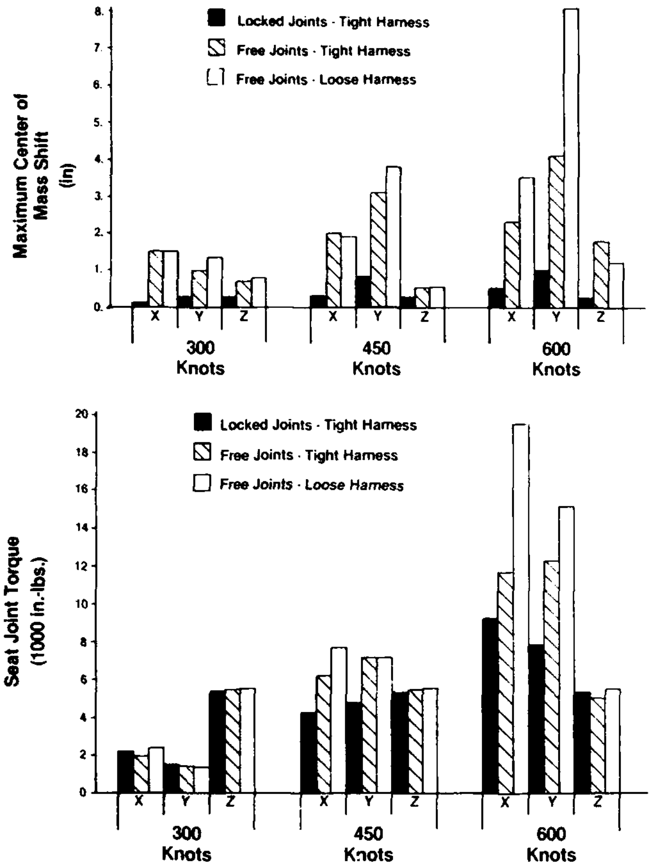


Fig. 11. Maximum center of mass shift and torque applied to the seat by the ATBM during the ejection simulated in Fig. 10. The 3 joint and harness conditions simulated illustrate the importance of biofidelity of joint properties in dummies used for ejection tests and the influence of harness tightness (9,19). (Upper) Center of mass shift. (Lower) Torque at the seat reference point.

for general use, many more details on human response to dynamic forces.

Our Articulated Total Body Model (ATBM) predicts body motion and deformation under complex multidirectional forces, such as escape acceleration and windblast (9). An ejection simulation is illustrated in Fig. 10. It gives us vital parameters for the advanced CREST control system design: the body and seat center of mass shifts and the torques exerted by the body on the seat (Fig. 11). The joint dynamics of the model, the elastic and resistive forces, are based on extensive measurements on human joints, particularly the complicated shoulder and hip complexes. The ejection simulations include different harness types and tightnesses. The model is used widely in automotive safety research, locomotion studies, and other applications (10). An example is shown in Fig. 12.

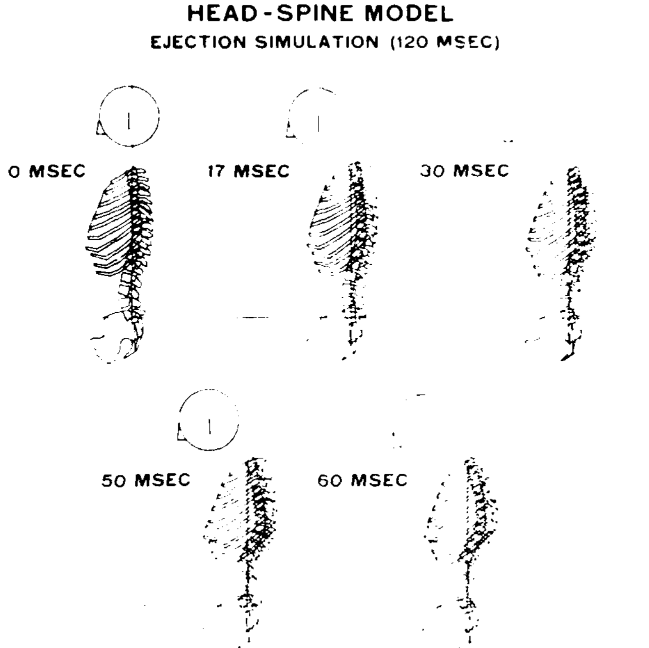


Fig. 13. Body deformation under ejection forces simulated with the head-spine model. Detailed motions and forces at each skeletal component are calculated.

The second model, which we are using now in many exploratory investigations, is the Head-Spine Model, which was derived to predict in detail the mechanical loads and injury probability along the spinal column. In other words, it analyzes in detail what is going on inside the simple spring of the DRI Model. In response to three-dimensional force inputs, it allows the calculation of the stresses at each vertebral level. Combined with the detailed failure characteristics of each vertebra discussed under biomechanical component research, it enables us to predict the probability of spinal injury at each point of the spine (Fig. 13,14) (19,20). It explains the different distribution, types, and severity of injuries collected in our accident data bank and helps prevent

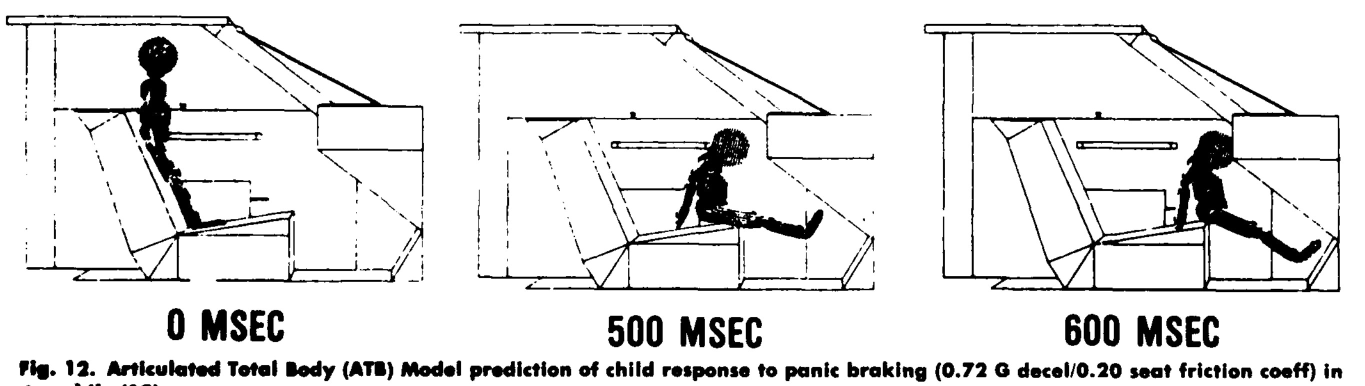


Fig. 12. Articulated Total Body (ATB) Model prediction of child response to panic braking (0.72 G decel/0.20 seat friction coeff) in automobile (10).

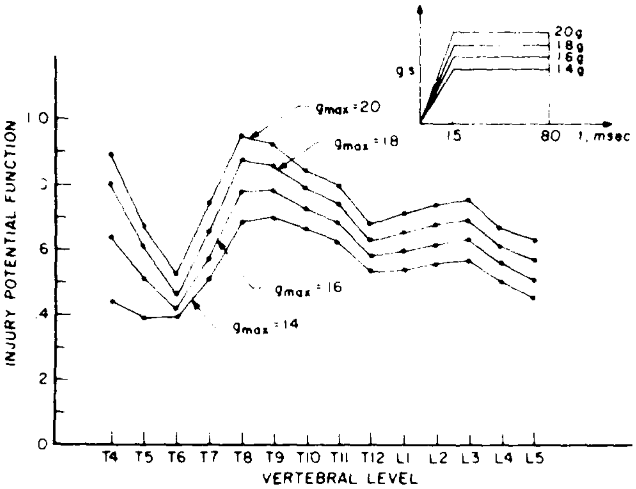


Fig. 14. Injury probability as a function of vertebral level predicted with the head-spine model (Fig. 13) for various ejection accelerations G_z. The injury potential function combines vertebral compression and torsion loads (9,19).

injurious stress concentrations in future designs. Fig. 15 illustrates the increase in predicted injury likelihood due to various head or helmet mounted devices (20). Inertial loads as well as attachment locations can markedly affect location and severity of the injury to be expected. Both models allow us to assess the effectiveness of various harnesses, seats, or seat-harness combinations, and to respond in all six DOFs, whereas the DRI concept separates the DOFs and incorporates a fixed optimal standard harness effectiveness.

I think it is only fair to conclude that the biomechanics research of the past 25 years did not lag behind the pioneering first steps of the previous decades. The payoff in terms of practical design guidance for aircraft escape systems and crashworthiness has been impressive. However, the payoff probably was still greater in terms of general knowledge accumulated, in that we acquired the capability to predict what will happen under load conditions and circumstances that were never experienced, but may arise in future missions. The space environment is only one example where our basic data

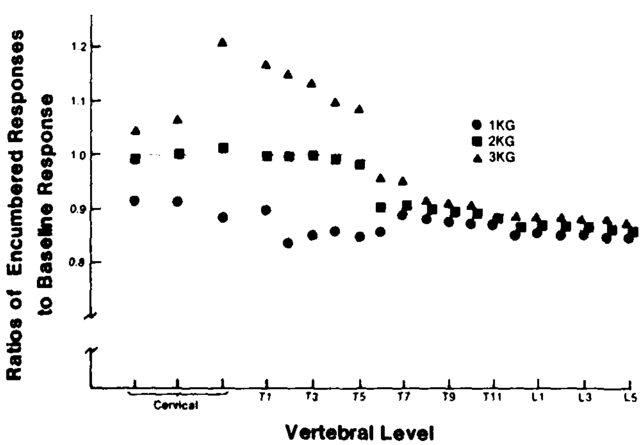


Fig. 15. Ratios of loading in the spine for 14 G_z exposures with 1, 2 and 3 kg head encumbrances to a baseline 17 G_z exposure with a 1 kg head encumbrance. Exposure accelerations used were half-sine profiles with 300 ms duration (20).

combined with our model predictions can help us predict perceptions and motion capabilities and can help us avoid potential mechanical hazards.

We hope that our accumulated biomechanical data and knowledge, component test data, human response and tolerance experiments, as well as accident and trauma analyses will all be collected in our Biodynamics Data Bank (1), which will grow to a national data bank and will be available to all scientists and engineers who can profit from it. As mentioned before, it is not only aerospace medicine and automotive crash research and protection which profit from these data, but also occupational medicine, sports medicine, ultrasonic imaging and other detection techniques, orthopedic biomechanics, and trauma research in general. I was only able in this short review to sketch some of the highlights of the developments and applications in the biodynamics field, concentrating on the contributions from our laboratory. Aerospace medicine was instrumental in opening up this new and exciting field of biodynamics and is still at its forefront.

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Although the active and passive strength of the human body has been of interest for many medical and ergonomic problems, it was the emergence of aerospace medicine which required the fundamental and practical studies that provide today's body of data in this field. After the early pioneering years of the 1930's, 1940's and 1950's, the last 25 years brought a maturing of the field of biodynamics, its methods, tools, and theoretical foundations. This overview discusses some of these advances, their contributions to aerospace safety, as well as their applications to the broader areas of traffic safety, orthopedic biodynamics, medicine and ergonomics. To meet future aviation as well as societal challenges, steady efforts by a few centers of excellence are required to integrate operational, experimental, and theoretical advances into sophisticated prediction capabilities.

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